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Feasibility analysis and design of a novel ventricular assist miniscule nutation pump

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Abstract

For the questions that the rotating continuously artificial heart pump would damage blood cells and cause hemolysis running at high rotating speed, this paper proposes a novel ventricular assist miniscule nutation pump based on nutation principle. The novel nutation pump aims to reduce the speed of the rotating parts under the premise of necessary flow rate and without the increasing of the pump volume. The equations for the pump flow discharge are obtained based on the mathematical modeling of pump flow discharge. The three-dimensional modeling of the novel ventricular assist miniscule nutation pump is further established. And the kinematic equations of the nutation disk are deduced, which provides the basis of the simulation. The flow field model of the pump is established and the meshing grids are divided in the software. Finally, the curve of volumetric flow rate, velocity vector diagram, pressure nephogram and shear stress nephogram in the internal pump are further obtained based on the simulation in the related software.

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Keywords: Artificial heart pump; Nutation pump; Nutation swing; Dynamic simulation.

1. Introduction

1.1 Application background

Heart is the original driving force of human blood system. Heart failure will disorder or disable the function of pumping blood, leading to the result that the blood flow is not enough to maintain normal metabolism of human body. Heart failure is not only known as one of the most serious illness causing death in the 21st century, but also the most challenging diseases [1]. It has shown that the global incidence of heart failure had continued to increase year by year. According to the survey, there have been 5 million heart failure patients in the United States with the growth of 10% every year [2].

Due to the amount of the heart donors is too small, and no effective immunosuppressive drugs are used for resolving the human tissue immune rejection after heart transplantation, it has become a trend that transplanting the heart with the artificial heart pump rather than the human body heart. Artificial heart pump is a device which promotes blood circulation by mechanical movement in order to completely or partly replaces human heart to pump blood.

1.2 Research background

With the characteristics of non-friction, small volume, high efficiency, long life, and so on [3], magnetic levitation heart pump becomes the new direction of artificial heart pump research [4].

However, almost all of the artificial heart pumps have hemolysis problems, which are mainly caused by high-speed rotating blade flapping blood. The volume is smaller and the efficiency of heart pump is higher, the working speed is higher. But the increased working speed of the pump blades causes more serious damage to blood cells. As the optimized design of blade surface has little effect on weakening the damage to blood cells, the magnetic suspension becomes new hotspots of research. However, if there is no change on the pump body, the high-speed problem of blades still exists.

For the aim to reduce the velocity of the rotating parts under the premise of necessary flow and no increasing volume, this paper proposes a novel miniscule nutation pump for the heart-assist. And a lot of work has been done, including the research on working principle, structure design, mathematical

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modeling of flow calculation and workflow field simulation.

2. The novel ventricular assist miniscule nutation pump

2.1 Basic principle

New miniscule nutation pump is based on the improvement of nutation motion shown in Fig. 1. For the nutation motion, the axis of a nutation disk has an angle with the space axis Z, and the disk rotates around the space axis with self rotating. Where in Fig. 1, when the intersection of the disk axis and the Z axis just lies on the center of the disk, the rotating speed of the disk is 0 and the disk is only doing nutation swing motion with its edge swing up and down. Then, the blood fluid separated by a baffle plate will form a circular directional flow motion and makes the pump sucking and discharging fluid.

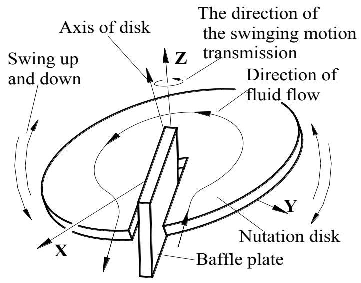


Fig. 1 The orientation of fluid flow

2.2 Structure design

The new miniscule nutation pump, shown in Fig. 2 (a), consists of pump body and drive motor. The pump body is composed of a upper cover, a sleeve, a nutation disk, a pump body shell, a lower cover, a flat clapboard, a cullis and lock screws as shown in Fig. 2 (b).

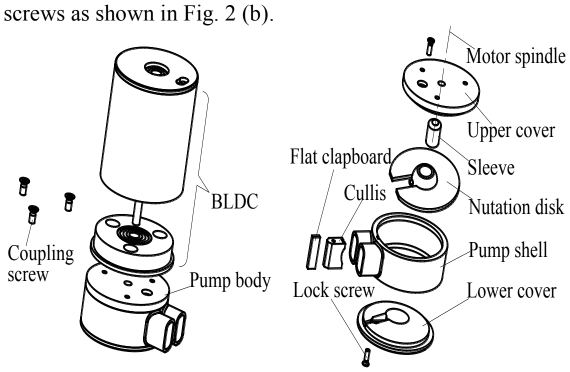


Fig. 2 (a) The component of pump; (b) Pump body assembly diagram

Where in Fig. 3, the motor shaft is fixed to the sleeve with incline axis (the tilt angle is nutation angle), and there is a through-hole in the spherical pair of the nutation disk. During the working time, the motor drives the sleeve rotation and the sleeve drives the nutation disk doing swing motion.

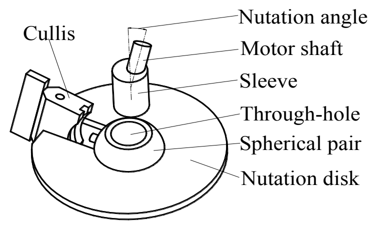


Fig. 3 Drive of sleeve

2.3 Working process

The internal vessel of pump is surrounded by a part of sphere and two inside cone. The nutation disk in tilt state within the vessel has contacting lines with the upper and under covers.

By taking the vessel which is under the nutation disk for an example as shown in Fig. 4, the under vessel is divided into two areas by the contacting line. The area connected to inlet is called inlet area and the area connected to outlet is called outlet area. When the motor shaft rotates counterclockwise, the nutation disk makes a cycle swing in counterclockwise. The contacting line also rotates counterclockwise, and the rotating speed of the contacting line is equal to the speed of the motor shaft. At this moment, the continuous increase area of the inlet area generates negative pressure, leading the fluid flow into the inlet area. Then the outlet area continuously diminishes and leads the fluid outflow from the outlet. When the under contacting line coincides with the position of cullis, the inlet area reaches the maximum value, and the outlet area reaches the minimum value. As shown in Fig. 4 (b), when the contacting line passes by the cullis, the outlet area changes from the minimum to maximum, and the change of inlet area is opposite. Then the above processes form a repeated working cycle.

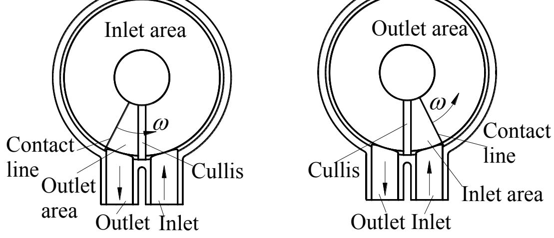


Fig. 4 (a) Contacting line not bypasses cullis; (b) Contacting line bypasses cullis

The working principle of above vessel is the same as under vessel. The motor drives the nutation disk making a counterclockwise swing, and drives the blood on either side of the nutation disk doing counterclockwise circular motion, forcing the fluid flow from the entrance (on the right) to the export (on the left).

3. Feasibility analysis for the nutation pump

3.1 Mathematical modeling of the pump flow discharge

According to the working principle of nutation pump, the pump belongs to positive displacement pump. Therefore, the working flow of pump can be obtained through calculating the rate of volume change of inlet area. By taking the under vessel as an calculating example shown in Fig. 5, the coordinate system $S_O(x, y, z)$ can be developed on the basis of designating the center of the vessel as the origin of coordinates O, the vertical direction as Z axis, the projection which the contacting line projected in the horizontal plane as X axis. The contacting line rotates around the Z axis clockwise. And the angular velocity ω, named as the motor speed, can be used for measuring the swing velocity of nutation disk.

Because the contacting line clockwise rotates around the Z axis, equivalently the gap of the under vessel counter-clockwise rotates around the Z axis, and the rotating velocity is ω. As shown in Fig. 5, the volume V_1 is the inlet area and the volume V_2 is the outlet area. At the t moment, the angle between the direction of gap and the X axis is $\theta(\theta = \omega t)$, and the relation between V_1 and time t is $V_1 = g(t)$. Thus the flow of the under vessel is $Q_1 = dg(t)/dt$.

As shown in Fig. 5, assuming that the angle rate of change $d\varphi$ of the position that has angle φ with the contacting line, and the infinitesimal volume corresponding to $d\varphi$ is dV_φ. Then V_1 can be obtained as

$$V_1 = \int_0^\theta dV_\varphi \tag{1}$$

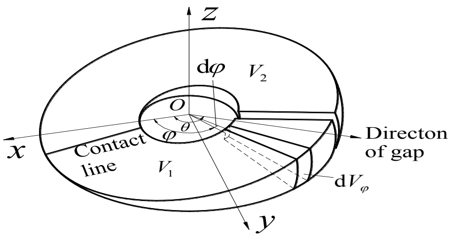


Fig. 5 The diagram of under vessel at t moment

As shown in Fig. 6, the infinitesimal volume dV_φ is separated out. The distance from origin O is r, and the angle with xoy plane is γ. The infinitesimal volume of the position is $dV_{\gamma r}$, and then $dV_{\gamma r}$ can be obtained as

$$dV_\varphi = \int_{R_1}^{R_2} \int_{-\alpha}^{\beta} dV_{\gamma r} \tag{2}$$

Where R_1 is the radius of spherical pair of nutation disk, R_2 is the radius of nutation plate edge. The upper and lower limits of γ is $-\alpha$ and β.

According to Fig. 6, the relation can be expressed as

$$dV_{\gamma r} \approx r d\gamma \cdot r d\kappa \cdot dr \tag{3}$$

and the projection of $d\kappa$ in the xoy plane is $d\varphi$, so there is $d\kappa = d\varphi \cdot \cos\gamma$, get into equation (3) have

$$dV_{\gamma r} \approx r d\gamma \cdot r\cos\gamma d\varphi \cdot dr \tag{4}$$

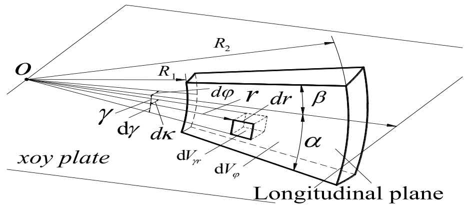


Fig. 6 Enlarge figure of infinitesimal volume

In order to obtain the function of β on φ, suppose that cylinder surface is cut off by a plane, which the dip angle is α, as shown in Fig. 7.

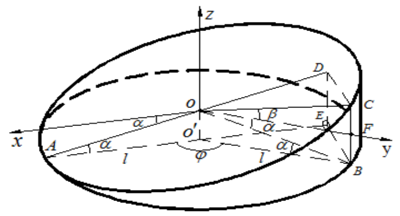


Fig. 7 Schematic diagram to solve β

In the ΔAED, there is

$$\overline{ED} = \overline{AE} \cdot \tan\alpha = (l - l\cos\varphi)\tan\alpha \tag{5}$$

$$\overline{BC} = \overline{ED} = (l - l\cos\varphi)\tan\alpha \tag{6}$$

$$\begin{aligned}\overline{CF} = \overline{BC} - \overline{BF} &= (l - l\cos\varphi)\tan\alpha - l\tan\alpha \\ &= -l\cos\varphi \cdot \tan\alpha\end{aligned} \tag{7}$$

so, the relative between β and φ is

$$\beta = \arctan\left(\overline{CF} \big/ \overline{OF}\right) = -\arctan(\cos\varphi \cdot \tan\alpha) \tag{8}$$

Thus, combining Equations (2), (4) and (8), there is

$$dV_\varphi = \int_{R_1}^{R_2} \int_{-\alpha}^{-\arctan(\cos\varphi \cdot \tan\alpha)} r^2 \cos\gamma d\varphi \cdot d\gamma \cdot dr \tag{9}$$

simplify the formula, there is

$$dV_\varphi = \frac{1}{3} \cdot (R_2^3 - R_1^3) \cdot \{\sin\alpha - \sin[\arctan(\tan\alpha \cdot \cos\varphi)]\} \cdot d\varphi \tag{10}$$

combining Equations (2) and (11), the volume equation of flow can be obtained

$$V_1 = \int_0^\theta \frac{1}{3} \cdot (R_2^3 - R_1^3) \cdot \{\sin\alpha - \sin[\arctan(\tan\alpha \cdot \cos\varphi)]\} \cdot d\varphi \tag{11}$$

And then the time derivative of volume of entrance area is equal to the flow, so

$$Q_1 = \frac{dV_1}{dt} = \frac{dV_1}{d\theta} \times \frac{d\theta}{dt} \tag{12}$$

where

$$\frac{dV_1}{d\theta} = \frac{1}{3} \cdot (R_2^3 - R_1^3) \cdot \{\sin\alpha - \sin[\arctan(\tan\alpha \cdot \cos\varphi)]\} \tag{13}$$

and there is $\theta = \omega t$, combining Eqs (12) and (13), the flow equation can be represented as

$$Q_1 = \frac{dV_1}{d\theta} \times \frac{d\theta}{dt} = \frac{1}{3} \cdot \omega \cdot (R_2^3 - R_1^3) \cdot \{\sin\alpha - \sin[\arctan(\tan\alpha \cdot \cos\varphi)]\} \quad (14)$$

Because the difference of contacting line between upper vessel and under vessel is 180° , and the flow of upper vessel is

$$Q_2 = \frac{1}{3} \cdot \omega \cdot (R_2^3 - R_1^3) \cdot \{\sin\alpha + \sin[\arctan(\tan\alpha \cdot \cos\varphi)]\} \quad (15)$$

3.2 The total capacity and the emissions of the pump

The total flow is the sum of flow of upper vessel and under vessel, i.e.

$$Q = Q_1 + Q_2 = \frac{2}{3}\omega(R_2^3 - R_1^3)\sin\alpha \quad (16)$$

The displacement of the pump P is the discharge of fluid volume when the contacting line rotates a circle. The period of contacting line rotation is $T = 2\pi/\omega$, then

$$P = QT = \frac{4\pi}{3}(R_2^3 - R_1^3)\sin\alpha \quad (17)$$

3.3 The determination of the parameters of the pump

As statistics shown, the radial size of most axial flow pump is ϕ25mm, the axial length is in the range of 55mm and 75mm, the work rotating speed is above 7500 r/min . Weighing the various aspects, the parameters of the pump can be set as the Table 1. shown.

Table 1. Parameters of novel ventricular assist miniscule nutation pump

Shell Diameter (mm)	Pump height (mm)	Radius of nutation pump R_2 (mm)	Spherical pair R_1 (mm)	Nutation angle (rad)
φ24	12	11	4	π/12

Therefore, combining the above parameters and Eq. (17), there is

$$P = \frac{4\pi}{3}\left(R_2^3 - R_1^3\right)\sin\alpha = \frac{4\pi}{3} \times \left(11^3 - 4^3\right) \times \sin\frac{\pi}{12} = 1373.6\left(\text{mm}^3\right) \quad (18)$$

The working flow requirement of heart pump is 5 l/min (namely $5 \times 10^6\ \text{mm}^3/\text{min}$), and the theoretical revolving speed is

$$n = \frac{Q}{P} = \frac{5 \times 10^6}{1373.6} = 3640(\text{r/min}) \quad (19)$$

As the calculating result shown, the work rotating speed of the novel ventricular assist miniscule nutation pump is lower in the premise of similar boundary dimension. So compared with other pump, the novel ventricular assist miniscule nutation pump has an advantage in protecting blood cells.

4. Characteristic Simulation and analysis

By utilizing the assembly model of the nutation pump built above, adding corresponding constraints to all the parts, the simulation for the nutation pump would be carried out.

4.1 Output flow curve of the pump

As the simulation result shown, when the velocity of nutation disk is 4000 times/min (namely that the speed of the motor is 4000 r/min), the rate of the pump output flow can be up to 83 cm^3/s (namely 5 l/min). It meets the flow requirements of heart pump [5]. As shown in Fig. 8, the flow curve is overall close to a straight line, which is consistent with the result of the theoretical calculation. The upper or under contacting line pass the gap of nutation disk at every half cycle and this makes the lack of the upper or under contacting line. Consequently, the temporary connection between the inlet and outlet areas occurs and that is the reason why the curve in Fig. 8 exits concave areas.

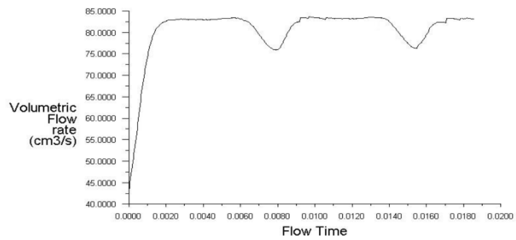
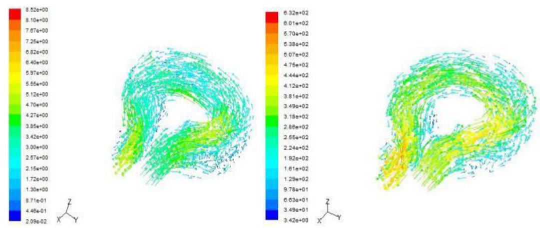


Fig. 8 The flow curve

4.2 Velocity distribution of pump

As shown in Fig. 9, the general trend of flowing fluid inside the pump is consistent. Because the flow cross section of the inlet and outlet are small, so the velocity of inlet and outlet are higher, and the velocity of rest are lower and more uniform. Tiny backflow appears in contacting line, because of the tiny gap between nutation disk and top or bottom cover cone. According to the diagram, flow dead zone does not exist in the internal flow field of pump, and it verifies that nutation pump has a antithrombotic property.

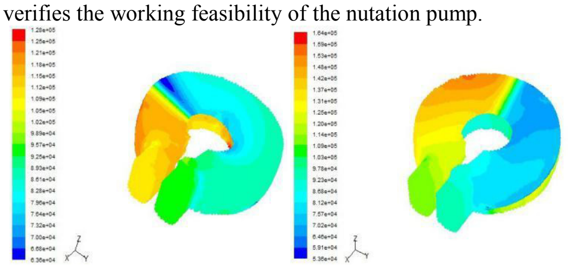


(a) t=0.25T (b) t=1T

Fig. 9 The velocity vector diagram

4.3 Pressure distribution inside pump

The pressure distribution inside the pump is shown in Fig. 10. What can be seen from the diagram is that there are two pressure distribution zones inside the pump, and the boundaries are the nutation disk and two pieces of contacting line. In the high pressure zone, pressure of the area that near the port of the lateral is higher, and the pressure of the inner area is lower. It is the result of centrifugal force in fluid circular motion. However, the pressure distribution of high and low pressure area is very uniform. The pressure of high pressure zone is slightly larger than the setting outlet pressure, which is 114658 Pa. And the pressure of low pressure zone is slightly less than the setting inlet pressure, which is 101325 Pa. The function of the connection of low pressure zone and inlet is "water absorption". And the function of the connection of high pressure zone and outlet is "flow extrusion". It also verifies the working feasibility of the nutation pump.

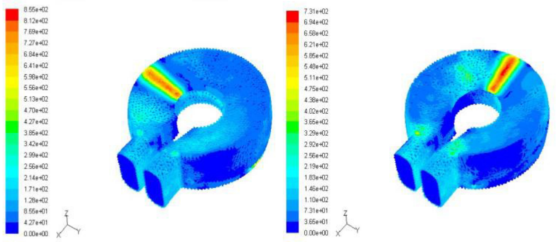


(a) t=0.25T (b) t=1T

Fig. 10 Nutation pump internal stress nephogram

4.4 Distribution of shear stress in the pump

It had been proven that the essential reason of hemolysis is causes by shear stress [6] and the shear stress can be divided into Newton shear stress (laminar shear stress) and Reynolds shear stress (turbulent shear stress), and the Reynolds shear stress is the main reason that cause hemolysis [6]. Therefore, the hemolysis performance of heart pump can be studied through the analysis of the distribution of shear stress. The simulation result for the distribution of shearing stress inside heart pump is shown in Fig. 11. Where in Fig. 11, the shear stress that appears at the position of the contacting line is higher than other areas, and the highest value is about 800 Pa. In addition, the shear stress at other area is very lower, almost under 150 Pa.



(a) t=0.25T (b) t=1T

Fig. 11 Internal shear stress of nutation pump

The hemolysis, because of blood cells rupture, is not only related to the flow field shear stress, but also associated with the time of red blood cells (RBC) exposure to shear field. Through experimental research, Niimi [7] obtained the relationship between shear stress and RBC exposure time, when hemolysis occurs. As shown in Fig. 12, there is a shear stress threshold, which is about 1000 Pa. When the shear stress exceeds the threshold, even if red blood cells expose very short time, blood cells damage can also occur. When the shear stress is between 150 Pa to 1000 Pa, red blood cell exposure time has a negative correlation with the shear stress. When the shear stress is less than 150 Pa, even if the exposure time of the red blood cells is infinite, red blood cells are not destroyed. According to the relationship between the exposures time of red blood cells and shear stress, except the position of contacting line, there isn't exist hemolysis at other area. However, though the shear stress of contacting line is larger than 150 Pa, the rotating speed of the contacting line is so fast that the contacting time is very short. So the red blood cells may not be broken.

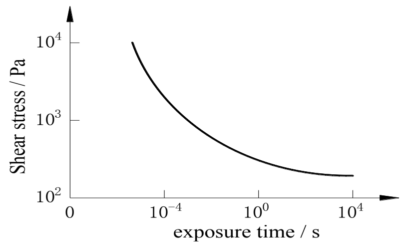


Fig. 12 The relationship between shear stress and exposure time of red blood cells

4.5 The import and export pressure change on the influence of the output flow

Different body posture will change artificial heart pump's pressure of the inlet and outlet, for example, the differential pressure between inlet and outlet at standing situation is higher than lying down. Therefore, a perfect artificial heart pump must be able to provide the same flow under different inlet and outlet pressure, or pump blood flow is not sensitive to pressure change of inlet and outlet. Due to the inlet and outlet of centrifugal rotary pump and axial rotary pump are always interlinked, the output flow is greatly affected by the pressure changes of the import and export, when the speed of blade rotation is constant. So the pump blood flow must be real-time monitored by sensors, and the speed of blade must be real-time adjusted, this is obviously increased the complexity and difficulty of control system.

In the simulation process, 5 groups differential pressures between the inlet and outlet are used to simulate. The speed of motor at all differential pressure is 4000 r/min, the pressure difference for inlet and outlet is respectively set to 6000 Pa, 8000 Pa, 10000 Pa, 12000 Pa and 14000 Pa. And the corresponding results of the output flow rate are 8, 82.84ml/s, 82.62ml/s, 82.39ml/s and 82.16ml/s. The relationship is given in Fig. 13.

As shown in Fig. 13, the relationship between output flow rate of nutation pump and pressure difference of the inlet and outlet is linear dependent. In the condition of motor speed is constant, the differential pressure of inlet and outlet rise 200 Pa, the output flow of nutation pump decreases about 0.23ml/s (0.014l/min). This suggests that, under different pressure, the output flow of nutation pump is very stable.

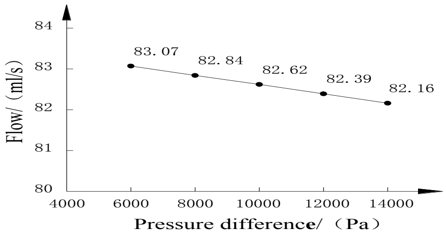


Fig. 13 The relationships between output flow rates of nutation pump and differential pressure of the import and export

The stability of the proposed pump can be also illustrated by other kinds of an axial-flow pump [8] shown in Fig. 14.

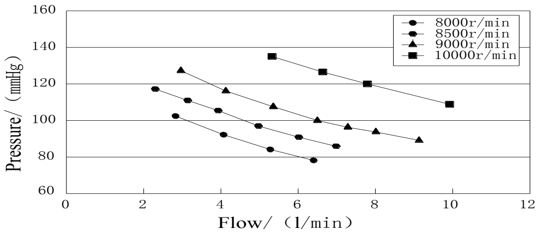


Fig. 14 The relationship between pressure of the inlet and outlet for an axial-flow pump

Where in Fig. 14, output flow rate of axial flow pump is and outlet pressure is tending to a straight line. The output pressure increases 20mmHg (about 2700 Pa), a decrease in the output flow rate is about 2.5l/min. According to the linear relationship, when the outlet pressure increasing 2000 Pa, the output flow rate of axial flow pump fell by about 1.85l/min. This value is far larger than the flow rate drop of proposed nutation pump.

5. Conclusions

This paper proposed a novel miniscule nutation pump for the heart-assist. The three-dimensional model is established. The working principle and working process of the new miniscule nutation pump are presented in detail. The mathematical modeling on flow discharge calculation is established and the flow discharge equation is obtained. According to the formula, the working flow of the pump has the linear relationship with the motor rotation speed and the sine relationship with the nutation angle, and has a negative correlation with radius of nutation disk edge. When the pump reaches the requirements of the output flow of heart pump, the rotating speed of motor is only the half of rotating speed of blades of the axial flow pump. The proposed miniscule nutation pump can reach the targets, including small volume, low speed and large flow rate.

From the results of simulation, the flow is consistent and can meet the demand of the flow of the blood. The stress field inside the pump can be separated into high and low pressure areas by the nutation disk, and the pressure of these two zones is uniformly distributed. The shear stress of pump fluid with $150\,pa$ is less than the critical value of producing hemolysis. During the working procedure, the output flow of miniscule nutation pump is stable.

Generally, the proposed novel nutation pump can achieve much lower velocity of the rotating parts under the premise of necessary flow without increasing its volume. The quality for protecting blood cells overmatches other kinds of existing heart pump. And without doubt, our further works will focus on the friction and wear reduction between the spherical bearing pairs and on the medical experiments.

Acknowledgements

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